

SPECIFICATION**TITLE****"DETECTION OF DIASTOLIC HEART FAILURE"****BACKGROUND OF THE INVENTION****5 Field of the Invention**

The present invention relates to an implantable medical apparatus for detecting diastolic heart failure (DHF), of the type having a DHF determining device for determining at least one blood pressure parameter for detecting a DHF state of the heart of a patient. The invention also relates to a pacemaker
10 provided with such an apparatus, and a method of detecting DHF, including the step of determining at least one blood pressure parameter for detecting a DHF state of the heart of a patient.

Description of the Prior Art

There is a growing recognition that congestive heart failure caused by
15 a predominant abnormality in the diastolic function, i.e. diastolic heart failure, DHF, is both common and causes significant morbidity and mortality. Therefore early detection of DHF is important. Patients do not, however, seem to have symptoms at an early stage. In addition it has been hard to separate diastolic and systolic heart failure and they may also exist
20 simultaneously.

It has been discovered that among the few parameters, separating diastolic heart failure from systolic heart failure, are certain blood pressure parameters obtained during work of the patient. Thus United States Patent No. 6,438,408 describes an implantable medical device for monitoring
25 congestive heart failure, CHF. A number of heart failure parameters indicative of the state of the heart failure are measured employing EGMs, blood pressures including absolute pressures, developed pressures (= systolic pressures – diastolic pressures) and the time derivative dP/dt , as well as heart chamber volumes. One of these parameters is the relaxation or contraction
30 time constant τ of the heart chamber. This constant τ is calculated from a

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continuous pressure signal and is the drop in ventricular pressure at the end of systole and in the first part of diastole. The τ parameter is thus a general parameter reflecting the relaxation process.

Patients suffering from DHF have both elevated ventricular filling pressure and elevated pulmonary venous pressure. To maintain cardiac output under load higher filling pressures are required to achieve adequate left ventricular filling, see H. Weinberger, Diagnosis and Treatment of Diastolic Heart Failure, Hospital Practice, <http://www.hosppract.com/issues/1999/03/weinb.htm>. However, for DHF

patients increased filling pressure does not result in increased left ventricular diastolic volumes. This is illustrated in figure 1 which shows results of studies on patients with isolated diastolic dysfunction, see D. W. Kitzman et al, "Exercise Intolerance in Patients with Heart Failure and Preserved Left Ventricular Systolic Function: Failure of the Frank-Starling Mechanism", J. Am. Coll. Cardiol., 17:1065-1072, 1991. Figure 1a shows the pulmonary capillary wedge pressure, PCWP, as a function of workload for patients with isolated diastolic dysfunction and for control persons not suffering for this dysfunction, and figure 1b shows corresponding curves for the left ventricular end-diastolic volume index, LVEDVI as a function of the workload. From the figures it appears that for patients suffering from the mentioned diastolic dysfunction, PCWP was elevated at rest and further elevated during exercise. This indicates that the ventricular filling pressure was increased under both conditions, whereas LVEDVI did not increase with exercise.

SUMMARY OF THE INVENTION

An object of the present invention is to utilize the above information to provide a technique for detecting DHF based on measurement of left atrial or pulmonary vein pressure.

Thus with the present invention the reduced peak and submaximal exercise performance of DHF patients are utilized for detecting DHF. With a sensor placed in left atrium or very close to the endocardium thereof, the left atrial pressure is measured during the diastasis phase just before atrial

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contraction. Alternatively a sensor in the pulmonary vein can be used, provided that pressure data for subsequent analysis are collected only when the pulmonary valve is closed. With the technique according to the invention it is possible to detect DHF at an early stage, often before the patient seems
5 to have any symptoms.

In the present invention the workload situation of the patient must be identified, and therefore, in an embodiment of the apparatus according to the invention, an activity sensor is provided for determining the workload of the patient.

10 Since the absolute pressure is used, measured pressure values must be corrected with e.g. the level of the mean pressure in the right atrium or in vena cava close to the right atrium. This correction is needed because any column of blood above the sensor location, which thus is posture depending, will add to the absolute pressure level. Such a correction will also
15 compensate for a varying ambient pressure. In an embodiment of the apparatus according to the invention a pressure compensator is therefore provided for correcting the measured absolute value of left atrial pressure with the mean (average) pressure in right atrium or in vena cava close to the right atrium.

20 In another embodiment of the apparatus according to the invention, the DHF determining device includes an impedance measuring unit for measuring the impedance between a left ventricular electrode and a second electrode intended to be positioned at a location so that most of variations in the measured impedance are due to varying left ventricular volume. ΔP values
25 and corresponding values of the difference, ΔV , in the end-diastolic ventricular volume, obtained from the impedances measured for the predetermined workload situation and for the patient in rest, are supplied to a quotient forming unit to form the quotient $\Delta P/\Delta V$, and a comparator compares this said quotient $\Delta P/\Delta V$ with a predetermined quotient reference value for DHF
30 detection. By adding information about left ventricular end-diastolic pressure the DHF detection can be improved. In a healthy person the end-diastolic volume increases with exercise but shows only a small increase in left atrial

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pressure. In a person suffering from DHF, who has a more stiff heart muscle, the left ventricular end-diastolic volume changes very little but there is a large increase in left atrial pressure. The quotient $\Delta P/\Delta V$ is consequently a useful quantity for improving the detection of DHF. The difference ΔV is determined

5 by impedance measurements.

In other embodiments of the apparatus according to the invention, a first averaging unit forms an average value of the absolute pressures measured during a of cardiac cycles with the predetermined workload situation and an average value of the absolute pressures measured during a
10 number of cardiac cycles with the patient in rest, and a second averaging unit forms an average value of the impedance measured during a number of cardiac cycles with the predetermined workload situation and an average value of the impedance measured during a number of cardiac cycles with the patient in rest. In this way the quality of the pressure measurements as well
15 as the impedance measurements are improved.

In other embodiments of the apparatus according to the invention a wireless communication unit is connected to the comparator for automatically sending the results of the comparisons with the reference values to an external receiver, or a memory is provided for storing the results of the
20 comparisons with the reference values. Thus if the measured absolute pressure, and possibly also the quotient $\Delta P/\Delta V$, have risen above the associated reference values in a predetermined way, this condition is automatically transmitted to a physician or stored for transmission in connection with a follow-up.

25 The invention also relates to a pacemaker provided with the apparatus for detecting DHF and a control unit that optimizes pacing therapy depending on the result of the comparisons with the predetermined reference values. The pressure measuring unit of the apparatus according to the invention then preferably is a pressure sensor connected to the pacemaker, since it can
30 monitor the pulse pressure of its carrier for long periods. This is an advantage since evolvement of DHF is a slow process.

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In an embodiment of the pacemaker according to the invention, a rate responsive sensor is used as an activity sensor for determining the workload situation of the patient. Even the pressure sensor of the pacemaker can be used as activity sensor.

5 In an embodiment of the method according to the invention, in a calibration procedure instantaneous values of the left ventricular volume are measured by an ultrasound technique and the relation is established between these measured ventricular volume values and simultaneously measured impedance values. Since the measured impedance depends on the location
10 of the electrodes used, it will vary from patient to patient and therefore such a calibration is required.

As mentioned above the measured difference between left atrial pressure measured in workload and rest situations is compared with a predetermined pressure difference reference value. The reference value for
15 detection of DHF is preferably obtained from measurements on the patient at an early stage of the implantation period of the apparatus or pacemaker. The patient is assumed not to suffer from DHF at the time of implantation. Therefore, in an embodiment of the method according to the invention, the pressures are measured for different workloads of the patient and for the
20 patient in rest at an early time, when the patient is not suffering from DHF, for determining the pressure difference reference value. Corresponding impedance measurements are preferably also made on the patient at this early stage for determination of the quotient reference value.

Figures 1a and 1b show results of studies on patients having a diastolic
25 dysfunction as explained above. From these figures it can be seen that a heart suffering from DHF has another PCWP versus workload behaviour than a normal heart, "Controls" in the figure. The present invention is based on this knowledge since the PCWP is approximately equal to the left atrial pressure LAP.

30 The detection of DHF by measurements of the absolute left atrial pressure LAP or the pressure in the pulmonary vein can be improved by adding information about left ventricular end-diastolic volume. In a healthy

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person the end-diastolic volume increases with workload or exercise, whereas the left atrial pressure shows only a small increase. In a person suffering from DHF the left ventricular end-diastolic volume changes vary little but there is a large increase in the left atrial pressure. This is illustrated by Figure 2, which is a diagram from D. W. Kitzman et al, "Exercise Intolerance in Patients with Heart Failure and Preserved Left Ventricular Systolic Function: Failure of the Frank-Starling Mechanism", J. Am. Coll. Cardiol., 17:1065-1072, 1991, plotting PCWP versus left ventricular end-diastolic volume. The open boxes in the figure indicate the direction of change from rest to exercise or workload for patients with DHF, and the solid boxes illustrate the corresponding direction of change for normal control persons. The shift upwards in the diagram for non-healthy patients should be noted.

The left ventricular end-diastolic volume is preferably determined by measurements of the electric impedance between e.g. a left ventricular electrode, normally placed in a coronary vein, and an electrode in right ventricle, right atrium or any other place in or on the heart, such that most of the variations in the measured impedance derive from variations in the left ventricular volume, as will be described below.

The detection and measurement of degree of DHF can be improved by studying the quotient $\Delta P/\Delta V$. ΔP denotes the difference between left atrial pressure measured in the workload and rest situations and ΔV the difference in the end-diastolic ventricular volume, obtained by the impedances measurements, for the predetermined workload situation and for the patient in rest, as explained above.

The quantity ΔV , and consequently the quotient $\Delta P/\Delta V$, is depending on the positions of the electrodes, used for the impedance measurements. These quantities will therefore vary from patient to patient. Thus some kind of calibration is needed.

The left ventricular volume varies during a cardiac cycle. The instantaneous volume can be measured by e.g. ultrasound technique. By simultaneously measuring corresponding impedance values a relation can be established between measured impedance values and left ventricular volume.

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The measured pressure difference ΔP can be directly obtained since the sensitivity of the pressure transducer is set by the manufacturer.

The reference value for the quotient $\Delta P/\Delta V$ for detection of DHF can be set from common knowledge about physiological properties during the
5 progression of DHF. In the same way the severity of the progression of DHF can be followed.

Figure 3 is a block diagram of an embodiment of a pacemaker according to the invention, including a pressure sensor 2, sensitive to the pressure in left atrium, LAP, of a patient's heart 4. The atmospheric pressure
10 and the pressure from blood volumes above the sensor also contribute to LAP. To compensate for this a sensor 6 is also provided for measuring the pressure in the right atrium or in vena cava. This compensation pressure is denoted by Pref.

A measurement value correlated to left ventricular volume changes ΔV
15 is obtained by measuring the impedance Z between an electrode 8 in or attached to the left ventricle and an electrode 10 in the right ventricle. The parameters in question are measured in the diastasis phase just before the atrial contribution to the left ventricular filling. For this purpose a timer can be used, starting at the preceding contraction. Alternatively P-waves can be
20 sensed by an electrode 12 implanted in the right atrium of the heart 4 for starting the measurement. Parameter values measured during a period preceding the atrial contraction are stored in a ring buffer. This procedure will be described below with reference to Figure 4.

The pacemaker in Figure 3 has ordinary circuits 14 for IEGM sensing
25 and stimulation. It also comprises circuitry 16 for measuring the impedance Z between implanted electrodes 8, 10 as mentioned above. To determine volume changes of heart chambers by impedance measurements between electrodes implanted in the heart is per se previously known from e.g. EP 1 384 492. Such impedance measuring circuitry is also known from respiration
30 activity measuring pacemakers.

The pacemaker illustrated in figure 3 also contains a unit 20 for measuring activity or workload of the patient by means of an activity or

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workload sensor 22. An ordinary microprocessor with supporting circuits 18 are provided for calculation and signal processing explained in the following with reference to Figures 4 and 5. Sensed Pref and LAP signals are digitized in A/D- converters 24 and 26 respectively before supplied to the
5 microprocessor and supporting circuits 18. Data related to sensing and pacing, the measured impedance Z, and the activity or workload Act of the patient are supplied to the microprocessor and its supporting circuits 18 as well.

For communication with and transfer of data to external equipment the
10 pacemaker is provided with a telemetry unit that is not shown in Figure 3.

Figure 4 illustrates the determination of the quotient $\Delta P/\Delta V$. Since DHF evolves slowly, necessary measurements may be performed regularly, preferably with long intervals, for a long time. A timer is therefore provided, block 28 in Figure 4, and the process illustrated in the figure, is started by a
15 check of status of the timer 28, block 30. The timer 28 may be set to e.g. hours, days or weeks.

As soon as the timer 28 has counted down to zero the measurements of LAP, Pref, the impedance Z corresponding to left ventricular volume, and the activity or workload Act of the patient are started, block 32 in Figure 4. It
20 should be noted that the impedance is inversely correlated to the volume, but since the impedance variations are small and superimposed on a larger absolute impedance, impedance measurements are anyhow suitable for this purpose.

Measured data are stored in a ring buffer with n positions and large
25 enough to store data from pre-atrial loading of the left ventricle, block 34

As soon as a P-wave is detected, block 36, stored values of LAP, Pref, Z and Act are collected from a memory position n samples before the point of the P-wave detection. The number n depends on several circumstances, like the shape of the P-wave, the filter in the P-wave amplifier, and the position of
30 the right atrial electrode 12. In this way data are obtained from a point of time immediately before a noticeable change of the ventricular volume caused by the atrial contribution.

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After this collection of parameter values average values of corresponding parameters, taken over several cardiac cycles, are created, block 38 in Figure 4 and block 40 in Figure 5. The average value creating process is illustrated in greater detail in Figure 5. The purpose of the average creation is to improve the quality of data.

The contribution from the atrium to the ventricular filling should not be part of the left ventricular volume measured in this procedure, as explained above. Thus data are extracted from a storage position n samples before the latest sample, block 42 in Figure 5, to avoid atrial contribution to the ventricular filling. Stored average values of LAP, Pref, Z and Act from previous heart beats are used, block 44, to produce a floating mean value by adding a fraction $1/k$ of the present value to the already existing "old" average value, according to the equation

$$\text{New_Average_Value} = \frac{\text{Value} + (k-1) \bullet \text{Old_Average_Value}}{k}$$

where "Value" means the present value.

This process is performed for each of the parameters LAP, Pref, Z and Act, block 46, and the new average values are stored, block 48 in Figure 5. Block 46 contains corresponding equations for creating new average values of each of the parameters LAP, Pref, Z and Act, "avg" denoting "average".

A return is then made, block 50, to the main procedure illustrated in Figure 3, where the next step consists in subtraction of Pref from LAP to obtain the corrected LAP value LAP_r as discussed above. Thus,

$$\text{LAP}_r = \text{LAP} - \text{Pref}$$

block 52 in Figure 4. The corrected value LAP_r is then used to create the quotient $\Delta P/\Delta V$, block 54, store it in a memory, block 55, and compare the quotient $\Delta P/\Delta V$ with a reference value for subsequent actions, block 56.

For this procedure the parameter values are preferably stored in an organized way, in this case in a table, see Table 1 below. This table is set up in a memory. The parameters are measured over time for different workload levels of the patient, and this is used to increase the number of available

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measurements to be used and to improve the results. When average values over several cardiac cycles of the parameters have been obtained LAP_r and Z are stored in a table according to how actual activity or workload level fits the activity or workload intervals in the table. The following Table 1 shows an example of how this can be done.

Activity interval	LAP_r	Z	Parameters set
Act_1 to Act_2	<i>Data</i>	<i>data</i>	1
Act_2 to Act_3	<i>no data</i>	<i>no data</i>	0
...
Act_{n-1} to Act_n	<i>Data</i>	<i>data</i>	1
N			Sum of above

Table 1

The table will contain very few data at the beginning of the use of the apparatus according to the invention. With every entry into this process data will be filled in the table, either directly or via an averaging process if data already exist in memory positions in question. An "1" in column "Parameters set" indicates that a true parameter has been obtained and stored in the table. An "0" indicates the opposite.

In Table 1 "N" denotes the number of activity or workload levels for which averaged parameter values have been stored and over which the quotient $\Delta P/\Delta V$ has to be determined.

For determining the quotient $\Delta P/\Delta V$, ΔP is first created for each indicated activity or workload interval in the table. All pressure data are referenced to the pressure for the patient in rest, P_{rest} . The corresponding process is performed for ΔV . Since N activity or workload intervals are indicated as active in Table 1, N – 1 quotients $K_i = \Delta P_i/\Delta V_i$ are obtained. The average value of the K_i – values is returned, and $K_{avg} = (\sum K_i)/(N-1)$ is obtained. Thus the quotient K_{avg} is calculated.

The quotient K_{avg} is then compared to a predetermined reference value, preferably established and stored at the beginning of the use of the apparatus according to the invention (when the patient possibly were not yet suffering from DHF). By storing subsequent quotients K_{avg} together with

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associated time marks the evolvement of the degree of DHF can be followed over time. A test can be set up to test whether the quotient has passed a certain limit for starting an automatic transmission of the DHF status of the patient to the physician. This message is transferred by means of the
5 telemetry link of the apparatus according to the invention.

Although modifications and changes may be suggested by those skilled in the art, it is the invention of the inventors to embody within the patent warranted hereon all changes and modifications as reasonably and properly come within the scope of their contribution to the art.

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